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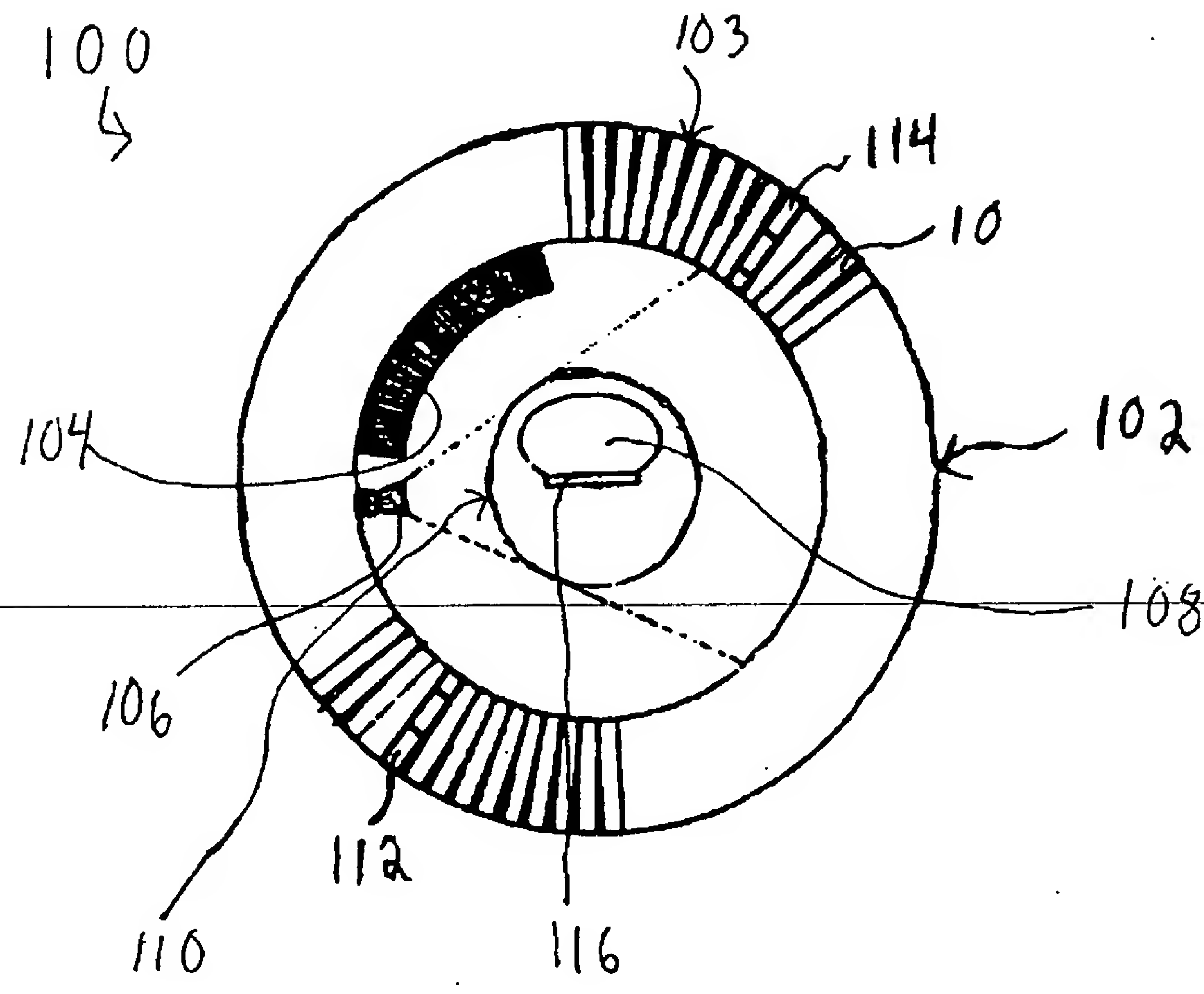
(54) **DETECTEUR POUR SCANNERS MULTIMODE**
(54) **DETECTOR ASSEMBLY FOR MULTI-MODALITY SCANNERS**



(57) A detector assembly for multi-modality PET/SPEC/CT scanners consists of a thin detector, on top of one or more deep detectors to detect low-energy gammas and high-energy annihilation photons. The deep detectors provides depth-of-interaction information for 511 keV detection in PET, while the thin front detector, that is essentially transparent to 511 keV and low-energy gammas, is used for detecting X-rays. Low-energy gammas can be detected by the thin detector on top and/or by the second deep detector of the assembly.

ABSTRACT OF THE DISCLOSURE

A detector assembly for multi-modality PET/SPEC/CT scanners consists of a thin detector, on top of one or more deep detectors to detect low-energy gammas and high-energy annihilation photons. The deep detectors provides depth-of-interaction information for 511 keV detection in PET, while the thin front detector, that is essentially transparent to 511 keV and low-energy gammas, is used for detecting X-rays. Low-energy gammas can be detected by the thin detector on top and/or by the second deep detector of the assembly.

TITLE OF THE INVENTION

DETECTOR ASSEMBLY FOR MULTI-MODALITY SCANNERS

FIELD OF THE INVENTION

The present invention relates to multi-modality scanners. More specifically, the present invention is concerned with a detector assembly for multi-modality scanners, in particular but not exclusively, an APD (Avalanche PhotoDiode) - based detector for multi-modality PET(Positron Emission Tomography)/SPECT(Single Photon Emission Computed Tomography)/CT(Computerized Tomography) scanners .

BACKGROUND OF THE INVENTION

The lack of anatomical information and the lack of information about the photon attenuation in the body in emission tomography imaging like SPECT (Single Photon Emission Computed Tomography) or PET (Positron Emission Tomography) imaging are major factors limiting the ability to accurately quantify radionuclide uptake in small regions of interest. Such lack of information about anatomical details and photon attenuation limits the diagnostic utility of emission imaging.

A drawback of emission tomography is that the spatial resolution obtained is limited. Typical emission tomographs have a resolution of the order of 5 to 15 mm. Another important drawback is that

the images produced are very noisy since the doses of radioisotopes that can be injected and the maximum counting rate are both limited. These two drawbacks render the delimitation of the regions of interest difficult.

Transmission imaging presents the advantage to have a sub-millimetric spatial resolution and thus allows to uncover anatomical details of the organ of interest. A drawback of transmission imaging is that it provides very little functional information.

The trivial solution to overcome the above mentioned drawbacks of each imaging method is to gather CT (transmission) and SPECT or PET (emission) images and to merge the anatomical and functional information. A way to achieve this is to collect the two sets of data using two different apparatuses, one gathering anatomical information and the other functional information, and to co-register one set with the other using sophisticated software.

A drawback of the latter method is that it can be difficult to superimpose the two sets of data since the measurements are done separately, with two different resolutions and most often with two different geometries. Moreover, movements of the patient as well as movements of the non-rigid structures within the body such as in the thorax or the abdomen add a blur to the resulting images.

The article entitled "The Design and Performance of a Simultaneous Transmission and Emission Tomography System", published in IEEE Transactions on Nuclear Science, Vol . 45, No 3, June 1998, and authored by Gullberg et al. proposes a simultaneous

transmission and emission tomography system. The system includes a detector to collect data from transmission and emission sources at different energies, and two other detectors to simultaneously acquire emission data.

A drawback of the system of Gullberg et al. is that the incorporation of a transmission-computed tomography system into a three-detector SPECT system causes problems of transmission data truncation and crosstalk between transmission and emission data windows. Another drawback relates to the used detectors which are limited in count rate and which cannot collect sufficient number of events in a short time for accurate anatomical definition of the body structures. Blur can still be caused by movements of the patient.

In United States Patent No. 5,376,795, issued on December 27, 1994, Hasegawa et al. describe an emission-transmission imaging system that uses a single detector for the transmission and emission data. The detector operates in a count or pulse mode to allow discrimination between the emission and transmission photons at low or medium count rates. At high count rate (with an X-ray tube, for instance) the detector can operate in a current mode without energy resolution. One advantage of this system is that both the transmission and the emission images are obtained by the same detector, and, thus, are intrinsically aligned. One drawback of the system of Hasegawa et al. is that it cannot be used to produce PET emission imaging.

OBJECTS OF THE INVENTION

An object of the present invention is therefore to provide a multi-modality scanner that possesses none of the above mentioned drawbacks of the prior art.

Another object of the invention is to provide a detector assembly for multi-modality PET/SPECT/CT scanners.

SUMMARY OF THE INVENTION

More specifically, in accordance with the present invention, there is provided a detector assembly for multi-modality scanners that consists of a scintillator transparent to high energy gammas and providing good detection for X-rays and low energy gammas and at least one scintillator for detecting higher energy gammas, and the circuits required to discriminate radiation on the basis of energy and crystal of interaction within the detector assembly.

In accordance with the present invention, there is also provided a detector assembly for multi-modality scanners that consists of a scintillator transparent to high and low energy gammas and providing good detection for X-rays and at least one scintillator for detecting low and high energy gammas, and the circuits required to discriminate radiation on the basis of energy and crystal of interaction within the detector.

According to a preferred embodiment of the present invention, the detector assembly for multi-modality scanners consists in a thin CsI(Tl) scintillator sitting on top of a deep LSO/GSO scintillator pair, read out by an avalanche photodiode (APD). The LSO/GSO pair provides depth-of-interaction information for 511 keV detection in PET, while the CsI(Tl), that is essentially transparent to 511 keV, is used for detecting lower energy gamma- and X- rays.

BRIEF DESCRIPTION OF THE DRAWINGS

In the appended drawings:

Figure 1 is a schematic view of a detector assembly for multi-modality scanners according to a preferred embodiment of the present invention;

Figure 2 is a schematic view of a multi-modality scanner using the detector assembly of Figure 1;

Figure 3 is a graph showing the total and PSD (Pulse Shape Discrimination)-gated energy spectra of 511 keV annihilation radiation taken with the detector assembly of Figure 1;

Figure 4 is a graph showing the total and energy-gated zero cross time spectra of 511 keV annihilation radiation taken with the detector assembly of Figure 1;

Figure 5 is a graph showing the total and PSD-gated energy spectra of ^{99m}Tc (140 keV gammas) taken with the detector assembly of Figure 1;

Figure 6 is a graph showing the total and energy-gated zero cross time spectra of ^{99m}Tc (140 keV gammas) taken with the detector assembly of Figure 1;

Figure 7 is a graph showing the total and PSD-gated energy spectra of ^{241}Am (60 keV X-rays) taken with the detector assembly of Figure 1; and

Figure 8 is a graph showing the energy spectra of ^{241}Am (60 keV) and ^{99m}Tc (140 keV) taken with the detector assembly of Figure 1 and illustrating the separation that can be obtained between X-rays and low-energy gammas in the front layer of CsI(Tl).

DESCRIPTION OF THE PREFERRED EMBODIMENT

A detector assembly 10 according to a preferred embodiment of the present invention will be described in connection with Figure 1 of the appended drawings.

The detector assembly 10 comprises a thin high-luminosity scintillator in the form a thin CsI(Tl) (Cesium Iodide (Thallium at a concentration of 10^{-3} on a per mole basis)) scintillator 12, a pair of

high-density scintillators 14 and a photodetector in the form of an avalanche photodiode (APD) 16.

CsI(Tl) scintillator 12 is used to detect X-rays and low-energy gamma-rays. Its thickness is chosen so as to absorb as completely as possible the X-rays used in CT and to absorb as little as possible of the high-energy 511 keV photons from the PET radiotracers.

The pair of high-density scintillators 14 includes a LSO scintillator 18 and a GSO scintillator 20. The LSO scintillator 18 is used to absorb as completely as possible the low-energy gammas (typically 140 keV) which are not completely absorbed in the CsI(Tl) scintillator 12 in the front layer. The LSO/GSO pair 14 is used for the detection of high energy photons (typically 511 keV) and so chosen as to discriminate scintillation decay times. CsI(Tl), LSO and GSO scintillators and avalanche photodiodes are believed to be well known in the art and will not be further described herein. Any other combination of suitable scintillators with the desired photon absorption and decay time characteristics can be used.

The GSO scintillator 20 is mounted on the APD 16 and the LSO scintillator 18 on the GSO scintillator 20, and the CsI(Tl) scintillator 12 is mounted on top of the LSO scintillator 18. The three scintillators 12, 18 and 20 are secured to one another and to the APD 16 by conventional means well known to those of ordinary skills in the art and are separated by optical grease.

The detector assembly 10 is configured to be used with the CsI(Tl) scintillator 12 facing the patient or other object on which measurement is made. The CsI(Tl) scintillator 12 is sufficiently thin to obtain a negligible efficiency of detection of high energy photons (typically 511 keV), but sufficiently thick to completely absorb low-energy X-rays and at least partially absorb medium-energy gammas. CsI(Tl) scintillator 12 is therefore relatively transparent to the high-energy annihilation photons of PET.

Also, the CsI(Tl) scintillator 12 has a scintillation decay time sufficiently different so as to be able to discriminate it from GSO and LSO by pulse shape discrimination methods.

Turning now to Figure 2, a multi-modality PET/SPECT/CT scanner 100, using detector assemblies such as detector assembly 10, will be described.

The scanner 100 comprises a gantry 102, provided with two diametrically opposite sets of detector assemblies 10, a collimator 104, an external source 106 to produce low energy radiation and a controller (not shown). Figure 2 illustrates a patient 108 who is investigated by means of the scanner 100. The scanner 100 is obviously provided with a frame or other means (not shown) with the gantry 102 mounted thereto.

Each of the detector assemblies 10 are fixedly mounted to the gantry 102 with the CsI(Tl) scintillator 12 turned inwardly to face the field of view 110, forming a detector ring 103. Each detector

assembly such as 112 of one set is positioned opposite to a detector assembly such as 114 of the second set relative to the field of view 110. The opposite pairs of detector assemblies such as 112 and 114 enable coincidence for PET imaging. Each detector assembly 10 is connected to processing electronics (not shown) used to amplify the signals from the detector and to discriminate detected events according to energy and pulse shape. The processing electronics are connected to the controller via electric cables (not shown) or other link means.

A collimator 104 is required for SPECT imaging. It is placed in front of the detector assemblies 10, facing the patient 108, to define preferential direction of incidence of the medium-energy photons (typically 140 keV) used for SPECT imaging. The collimator 104 can be a section of arc, as shown in Figure 2. It is then rotated inside the detector ring 103 to obtain projections over different incidence angles sequentially, using only a fraction of the detectors over the ring. In such case, the remaining detectors over the ring which are not used for SPECT imaging can be used for PET and/or CT imaging simultaneously. Alternatively, the collimator can also form a fixed full ring that fits inside the detector ring 103 to perform SPECT imaging using all the detectors at the same time. In either case, the collimator 104 must be retractable axially from inside the ring of detectors to permit PET or CT imaging with the full ring of detectors when SPECT imaging is not used.

The external source 106 is used for CT imaging. It can be in the form of an X-ray tube or of a high-intensity radionuclide source emitting low-energy photons (X-rays or low-energy gammas). As illustrated in Figure 2, the source 106 is mounted on the inner face of the

detector ring 103. Alternatively the source 106 can be embedded or incorporated into the structure of the gantry 102. In that case, the gantry 102 must be further provided with an inner aperture to enable projection of the X-rays in the direction of the patient 108. In either case, a mechanism (not shown) is provided to rotate the source 106 around the patient in order to obtain projection data over all incidence angles for CT image acquisition. X-ray tubes are believed to be well known in the art and will not be further described herein. The X-ray tube 106 is connected via electric cables (not shown) or other link means to the controller.

High-intensity, low-energy radionuclide source, and the associated shielding, collimation and shutter mechanisms that can be used for CT imaging are also well known in the art and will not be further described herein.

The controller can be a computer, or other electronic devices, provided with circuitries and/or programs to gather, discriminate and characterize the radiation-detector interactions. A display can be connected to the controller to show the resulting images. Construction of the controller and display is believed to be within the ability of one skilled in the art and, accordingly, will not be further discussed.

In operation, a patient 108 is first positioned on a table 116 situated approximately in the center of the gantry 102. Restraining means such as straps (not shown) can be used to immobilize the patient 108 on the table 116. The patient 108 has been previously injected with a small amount of radiotracer that targets the tissue or the organ of

interest. The radiotracer can contain two types of radioisotope: one for the PET imaging and the other for the SPECT.

The PET radiotracer is producing high-energy gamma photons that are emitted in pairs colinearly in opposite direction. The energy of each photon is approximately 511 keV. They are detected in coincidence by two diametrically opposite detector assemblies 10 (see for example detector assemblies 112 and 114). These photons pass through the CsI(Tl) scintillators 12 that are substantially transparent to these high energy photons. The photons are then detected by the LSO/GSO pairs 14. A signal is then sent to the controller that includes a coincidence circuitry (not shown). This signal is characterized by the pulse amplitude, the scintillation decay time and by the depth-of-interaction. This information is used to discriminate the type of interaction and its origin. The depth-of-interaction is also used to increase the spatial resolution of the image that will be produced by the controller.

The radiotracer can also produce mono-photonic gammas for the SPECT imaging . The energy of these photons is lower than the energy of the PET photons and generally higher or similar to the energy of the CT photons. The direction of propagation of the photons is determined by the collimator 104. The collimator 104 is configured taking into consideration the energy of the radiotracer and the type of measurement.

A SPECT photon can be detected by both the CsI(Tl) scintillator 12 and by the LSO 18 of the LSO/GSO pair 14. The signals

produced by the detector assembly 10 are characterized by the pulse amplitude and the scintillation decay time.

The CsI(Tl) scintillator 12 also detects photons coming from the external source 106 after crossing the region of interest without being absorbed or scattered through a large angle. This type of interaction is believed to be well known to those of ordinary skill in the art and, accordingly, will not be described in further detail.

For the CT imaging, the detector assemblies 10 are either in integration mode or in counting mode. To limit the irradiation of the patient 108 and the background noise, the external source 106 can also be collimated in the direction of detector assemblies 10. Likewise, to limit the background noise, especially the one due to scatter radiation from within the patient 108, a fan beam or cone beam collimator with focal point at the source position can be used in front of the detectors. In another embodiment of the present invention, the same collimator used for SPECT imaging having the appropriate focal length can also be used for CT.

As well known to those of ordinary skill in the art, CsI(Tl) signals can be easily discriminated from the LSO/GSO signals by classical pulse shape discrimination techniques.

It should be noted that, the scintillation signals from the three types of scintillators of the detector assembly 10, are converted to electric currents by the APD, the electric currents are converted to voltage pulses by the front-end analog electronic circuitry (not shown), and the

voltage pulses are further analyzed and converted into digital signals by the front-end processing electronic circuitry before being sent to the controller.

One of the advantages of the detector assembly 10 according to this invention is its discrimination properties that allow a same detector assembly 10 to gather simultaneously PET, SPECT and CT images.

Experiments have been conducted to show these discrimination properties. The tested detector assemblies were similar to the detector assembly 10 of Figure 1.

A 5x5 mm^2 active area reverse avalanche photodiode operated at an internal gain of 125 was used throughout this experiment. All crystals of the scintillators were polished on all faces and had a 4x4 mm^2 cross section. A 3 mm thick CsI(Tl) crystal was placed on top of an LSO and a GSO crystal, both having 10 mm in length. These were all coupled together with optical grease and wrapped in several layers of Teflon tape. The assembly was mounted on the APD through the GSO end also using optical grease. Signals from the APD were collected by a charge sensitive preamplifier, shaped with a spectroscopic amplifier having a shaping time of 250 ns and processed with a zero cross time PSD (Pulse Shape Discrimination) circuit. Measurements were performed using radioactive sources of ^{68}Ge (511 keV), ^{99m}Tc (140 keV) and ^{241}Am (60 keV).

The performance of the APD-GSO/LSO/CsI(Tl) detector is demonstrated in Figs. 3 to 8 for side irradiation of the crystal assembly by ^{68}Ge, ^{99m}Tc and ^{241}Am. The most salient features of the detector are summarized in Table I.

Table I - Results of pulse height (PHD) and pulse shape discrimination (PSD) measurements.

		GSO/LSO/CsI(Tl)		
		GSO	LSO	CsI(Tl)
511 keV	ΔE (%)	14	13	9.5
	Δt_{PSD} (ns)	10	4	18
	Zero Cross (ns)	38	24	130
140 keV	ΔE (%)	22	21	14
	Δt_{PSD} (ns)	-	~15	38
	Zero Cross (ns)	38	24	130
60 keV	ΔE (%)	-	31	21
	Δt_{PSD} (ns)	-	~20	64
	Zero Cross (ns)	-*	~20*	102

* PSD time peaks overlap.

Table I and Figures 3 to 8 show that the CsI(Tl) scintillator can be easily discriminated and yields suitable energy

resolution in an APD-based detector assembly designed for PET/SPECT/CT multi-modality imaging.

Table I and Figures 3 to 6 show that the LSO scintillator can be easily discriminated from GSO and CsI(Tl) at 511 keV and from CsI(Tl) at 140 keV. The fact that LSO is not fully discriminated from GSO at 140 keV is not harmful since 140 keV photons are very unlikely to reach the GSO scintillator located at the back of the detector assembly after crossing the front layers of CsI(Tl) and LSO. Similarly, the fact that LSO can be merely discriminated from CsI(Tl) at 60 keV has no consequence since X-rays are not expected to reach LSO behind the front layer of CsI(Tl). The LSO scintillator is shown to yield suitable energy resolution at 511 keV and acceptable energy resolution at 140 keV in an APD-based detector assembly designed for PET/SPECT/CT multi-modality imaging.

Table I and Figures 3 and 4 show that the GSO scintillator can be easily discriminated from LSO and CsI(Tl) at 511 keV and that it yields suitable energy resolution at 511 in an APD-based detector assembly designed for PET/SPECT/CT multi-modality imaging.

Alternatively, the external source 106 can be a radioactive element having a high activity. Also, a pair of LSOs, with short and long decay times, could be used instead of the LSO/GSO pair 14.

The separation between the scintillators of the same detector assembly 10 can be accomplished by shape discrimination of the scintillation rise time of each scintillator 12, 18 or 20, by a combination of shape and amplitude discrimination of the scintillation pulse on each scintillator, or by individual readouts of each scintillator 12, 18 or 20.

Also, any detector with suitable properties for detection of X-rays at high rate can be used instead of a scintillator. Similarly, detectors suitable for detection of low-energy gammas and high-energy annihilation photons could be used instead of the LSO/GSO pair 14. Special readout means, electrooptical or electronic, would then be used to collect the signals from the detectors.

The detector assembly 10 or the controller can include an apparatus or a method to detect the depth of the interaction of high energy photons.

Each detector assembly 10 represents a picture element (pixel) providing the multi-modality PET/SPECT/CT scanner 100 with a high counting rate. The detector assemblies 10 which represent pixels can be arranged into 1-D or 2-D close-packed arrays with individual readout and circuitry for independent processing of the signals from each detector pixel. One significant advantage of the detector assembly 10 when used in combination within large detector arrays is the very high overall count rate capability of the system, since the signals from every individual detector assembly 10 are processed in parallel by independent circuits.

Alternatively, small areas of the array of pixel detectors can share common front-end circuitry by combining the signals from a few detector assemblies (typically 4) using Anger-type coding logic to assign detected events to pixels. Such coding schemes are well known to those skilled in the art and will not be described further here. One advantage of such coding is a reduction of the number and density of electronic channels needed to process the signals from the individual pixel detector assemblies. However, one drawback is the increased dead time and reduced count rate per surface area achievable, and thus the limited contrast that can be obtained in CT images.

Merging of the anatomical and functional images is automatic since both modalities share the same detector geometry and are intrinsically aligned.

The CT information can thus be used independently or for attenuation correction of the emission imaging (PET/SPECT).

Although the present invention has been described hereinabove by way of preferred embodiments thereof, it can be modified, without departing from the spirit and nature of the subject invention.

WHAT IS CLAIMED IS:

1. A detector assembly for multi-modality scanners for detecting low and/or high energy radiation emitted from a source under investigation, said detector assembly comprising:

a low energy radiation detector substantially transparent to high energy radiation but responsive to low energy radiation from the source, said low energy radiation detector producing, in response to said low energy radiation, first radiation characterizing signals; and

a high energy radiation detector located downstream the low energy radiation detector and responsive to high energy radiation from the source, said high energy radiation detector producing, in response to said high energy radiation, second radiation characterizing signals.

2. A detector assembly as recited in claim 1, wherein at least one of said low and high energy radiation detectors comprises a scintillator.

3. A detector assembly as recited in claim 2, further comprising a photodetector optically coupled to said scintillator, said photodetector producing, in response to the radiation characterizing signals from said scintillator, corresponding electric signals.

4. A detector assembly as recited in claim 3, wherein the photodetector is selected form the group consisting of an avalanche photodiode, a pin diode and a photomultiplier tube.

5. A detector assembly as recited in claim 3, wherein the low energy radiation detector comprises a high-luminosity scintillator.

6. A detector assembly as recited in claim 5, wherein said high-luminosity scintillator is a CsI(Tl) scintillator.

7. A detector assembly as recited in claim 1, wherein said high energy radiation detector includes at least one high-density scintillator.

8. A detector assembly as recited in claim 7, wherein said high-density scintillator comprises a LSO scintillator.

9. A detector assembly as recited in claim 7, wherein said high-density scintillator comprises a GSO scintillator.

10. A detector assembly as recited in claim 6, wherein said high energy radiation detector includes:

a LSO scintillator downstream of and optically coupled to the CsI(Tl) scintillator; and

a GSO scintillator downstream of and optically coupled to the LSO scintillator.

11. A detector assembly as recited in claim 1, wherein the low energy radiation includes X-rays and said high energy radiation includes high energy gammas.

12. A detector assembly as recited in claim 11, wherein said low energy radiation further includes low energy gammas.

13. A detector assembly as recited in claim 11, wherein said high energy radiation further includes low energy gammas.

14. A multi-layer detector assembly for multi-modality scanners for detecting X-rays and/or high energy gammas emitted from a source under investigation, said detector assembly comprising:

a first X-ray detecting layer substantially transparent to high energy gammas and producing, in response to X-rays from the source, X-ray characterizing signals; and

a second high energy gamma detecting layer located downstream the first layer and producing, in response to high energy gammas from the source, gamma characterizing signals.

15. A multi-layer detector assembly as recited in claim 14, wherein at least one of said first and second layers is a scintillator and the characterizing signals produced by the scintillator include scintillation light, and wherein said multi-layer detector assembly further comprises a photodetector layer which produces, in response to the visible light, a corresponding electric signal.

16. A multi-modality scanner comprising:

a gantry having a longitudinal axis;

a scanner operation controller;

at least one pair of diametrically opposite detector assemblies as recited in claim 1, the detector assemblies of said pair mutually facing each other and being connected to the scanner controller whereby the first and second radiation characterizing signals from said detector assemblies are supplied to said scanner operation controller;

a low energy gamma radiation source movably mounted to the gantry, and connected to and controlled by the scanner operation controller; and

a collimator mounted rotatable to said gantry about the longitudinal axis, said collimator being connected to and controlled by said scanner operation controller.

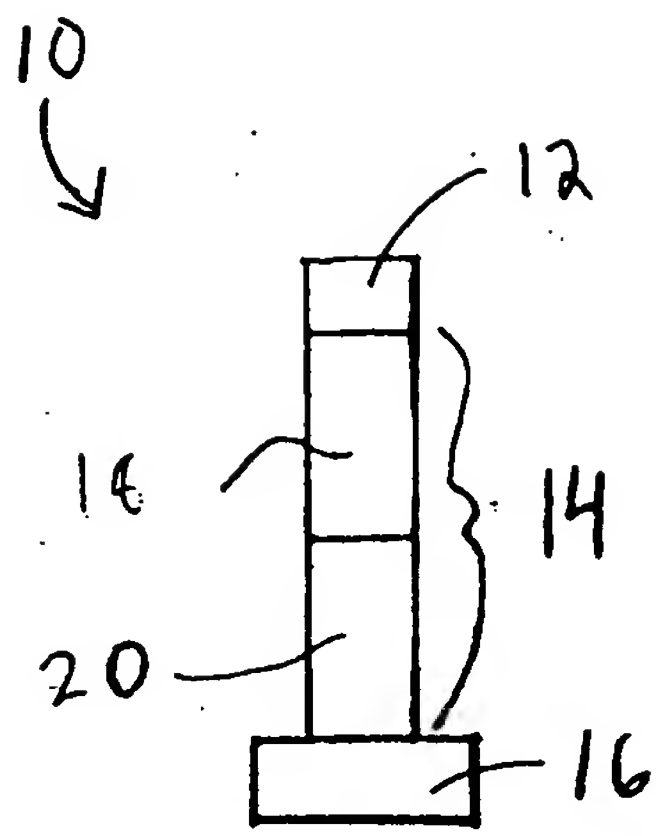


FIG. 1

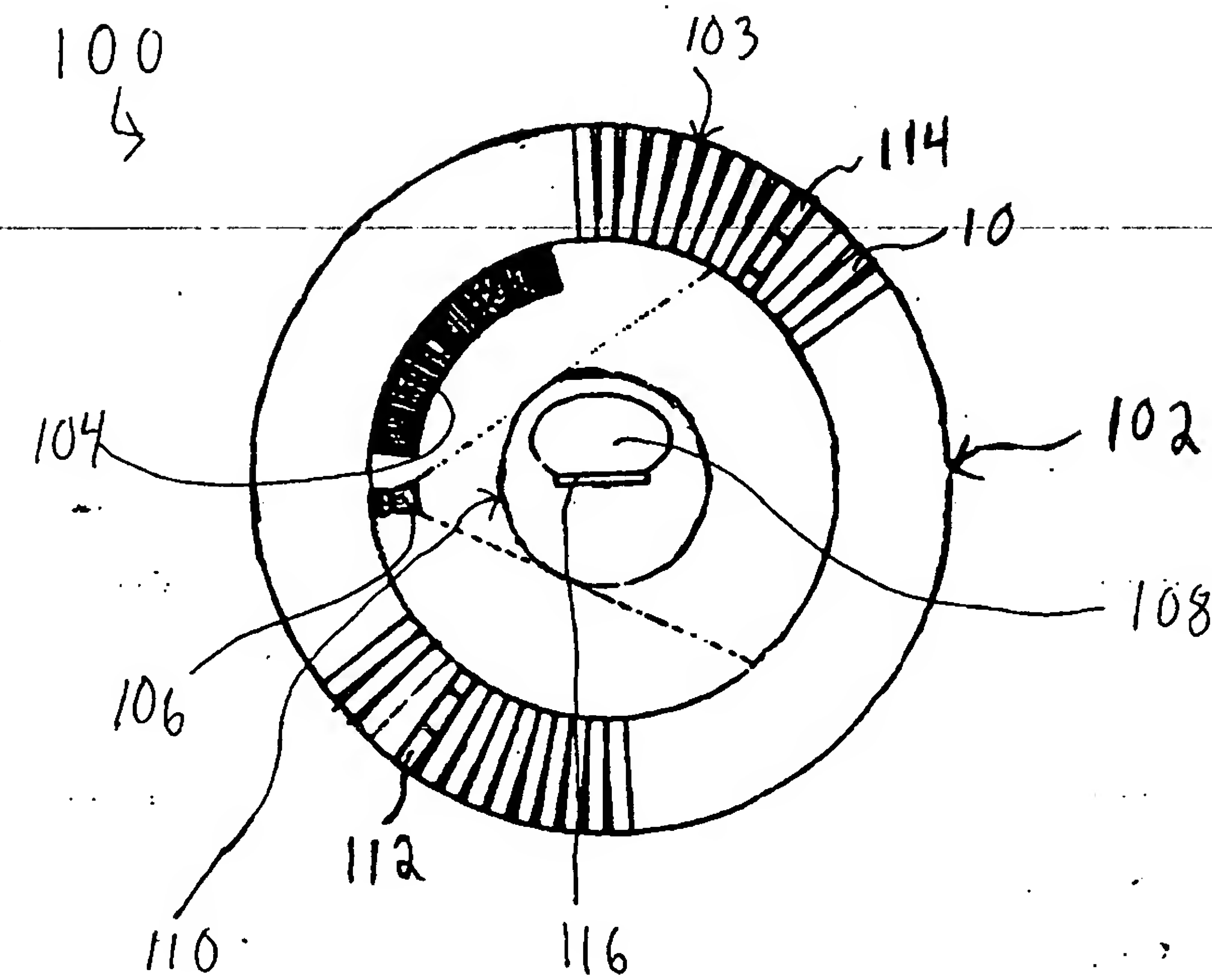


FIG. 2

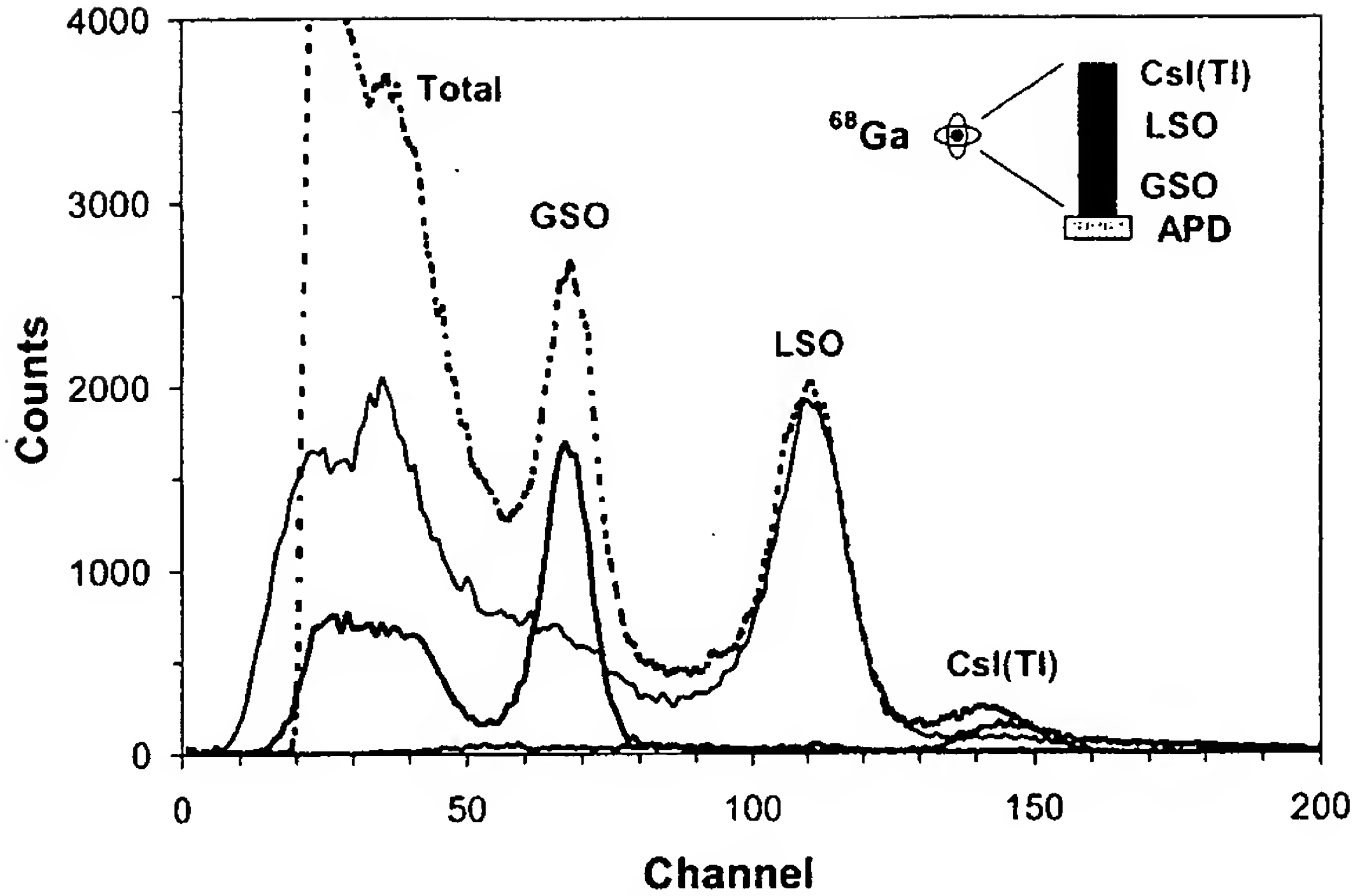


Fig. 3 – Total and PSD-gated energy spectra of ^{68}Ga (511 keV) taken with the detector assembly of Figure 1.

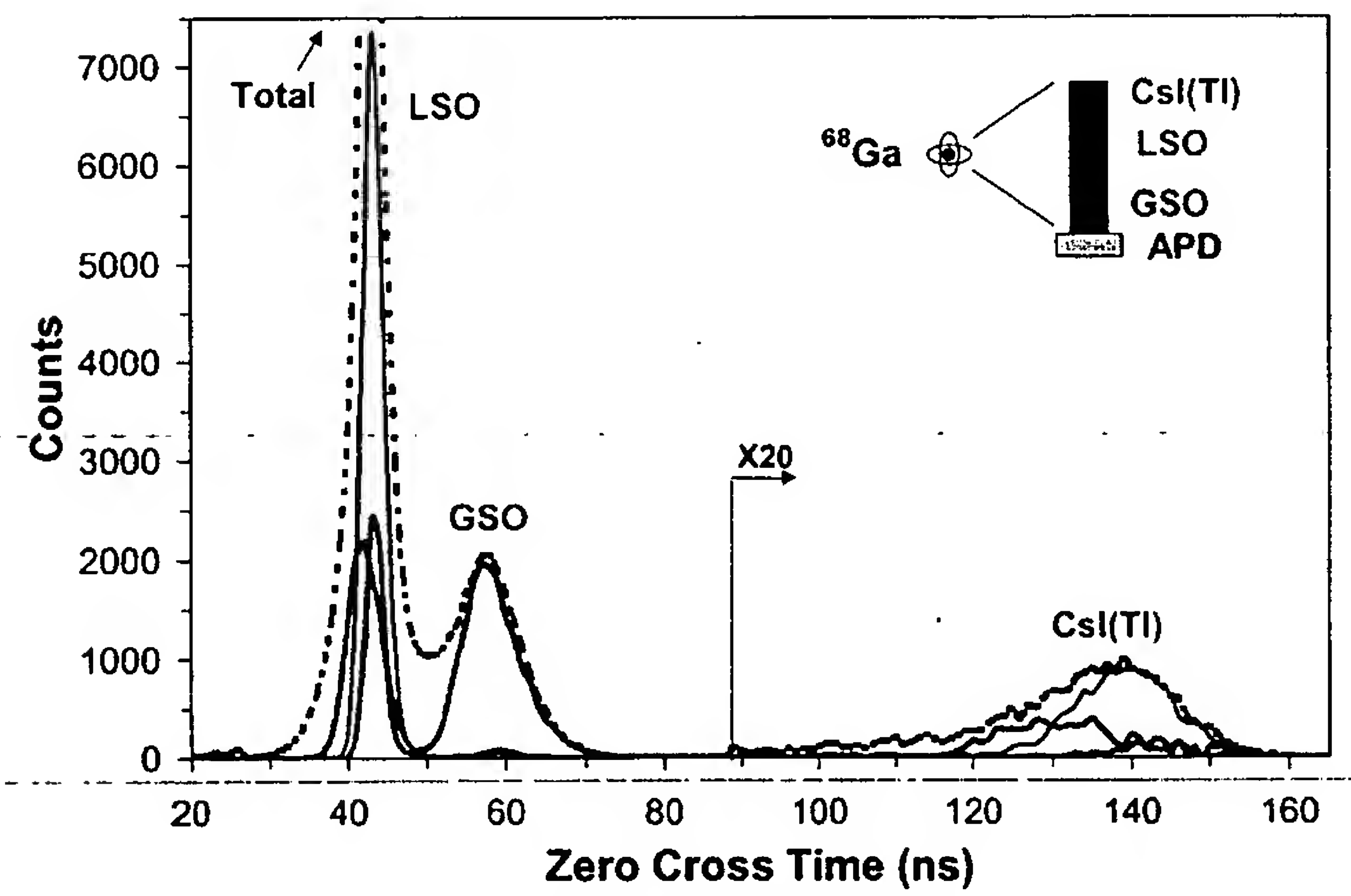


Fig. 4 – Total and energy-gated zero cross time spectra of ^{68}Ga (511 keV) taken with the detector assembly of Figure 1.

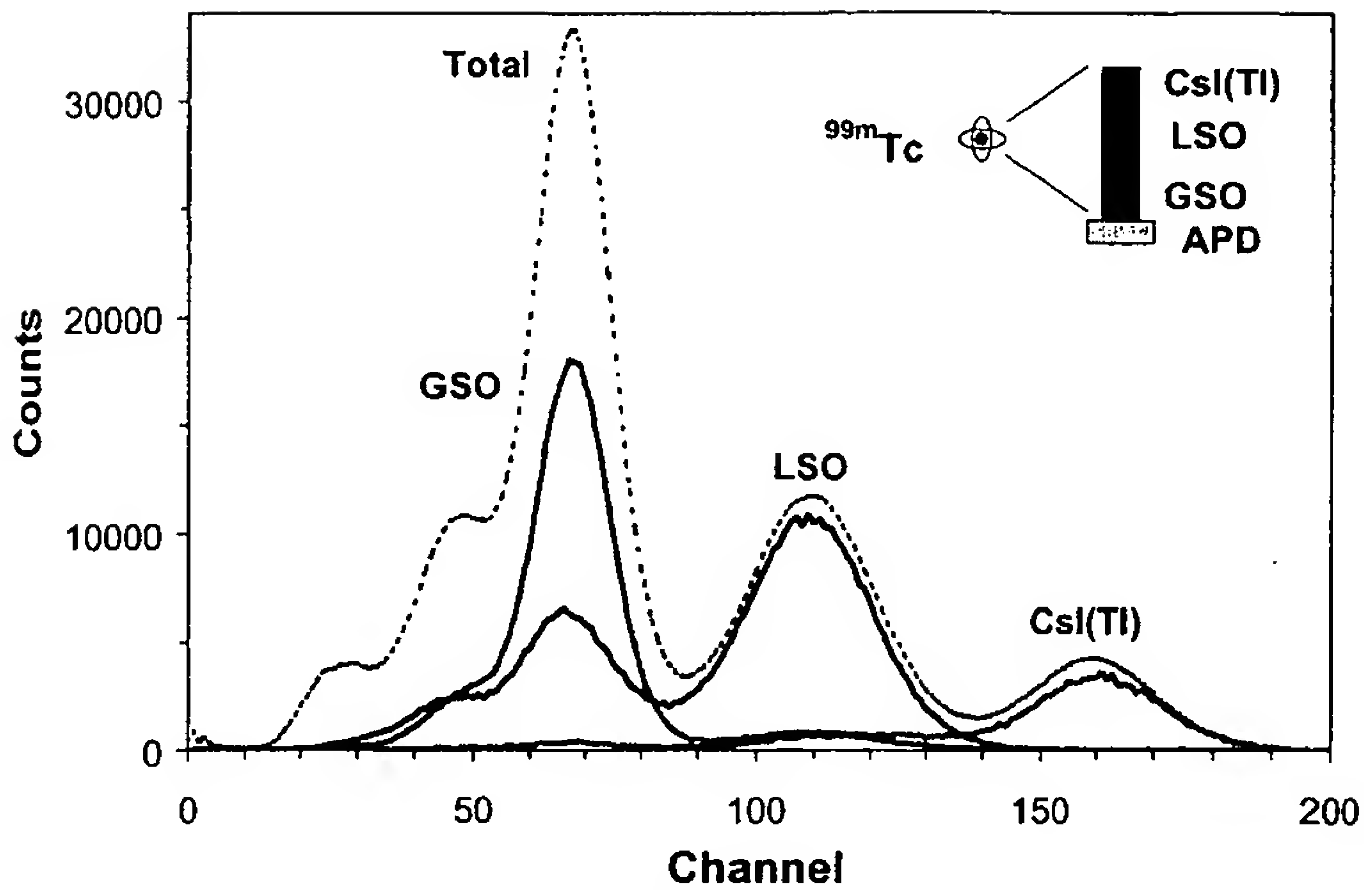


Fig. 5 – Total and PSD-gated energy spectra of ^{99m}Tc (140 keV) taken with the detector assembly of Figure 1.

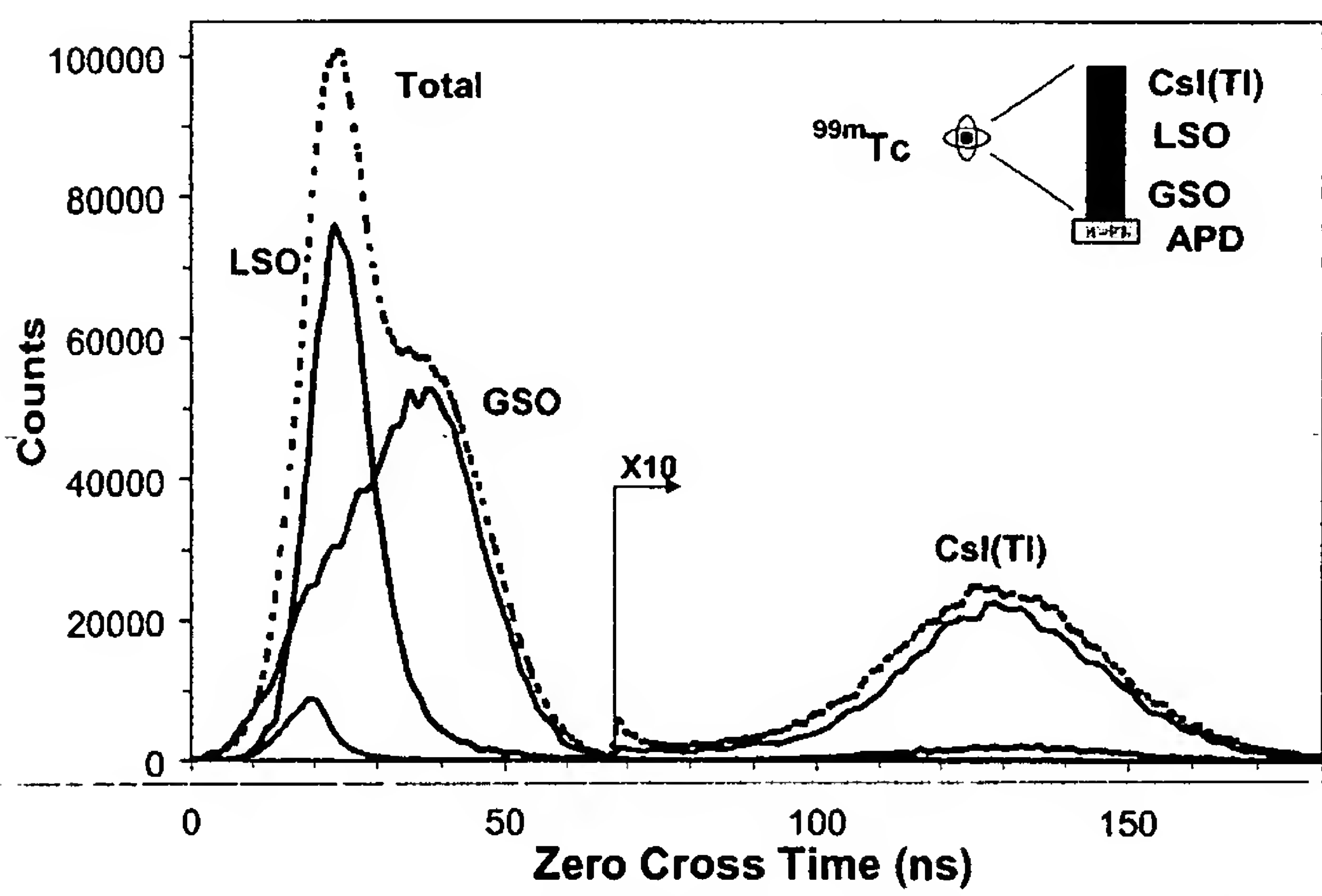


Fig. 6 – Total and energy-gated zero cross time spectra of ^{99m}Tc (140 keV) taken with the detector assembly of Figure 1.

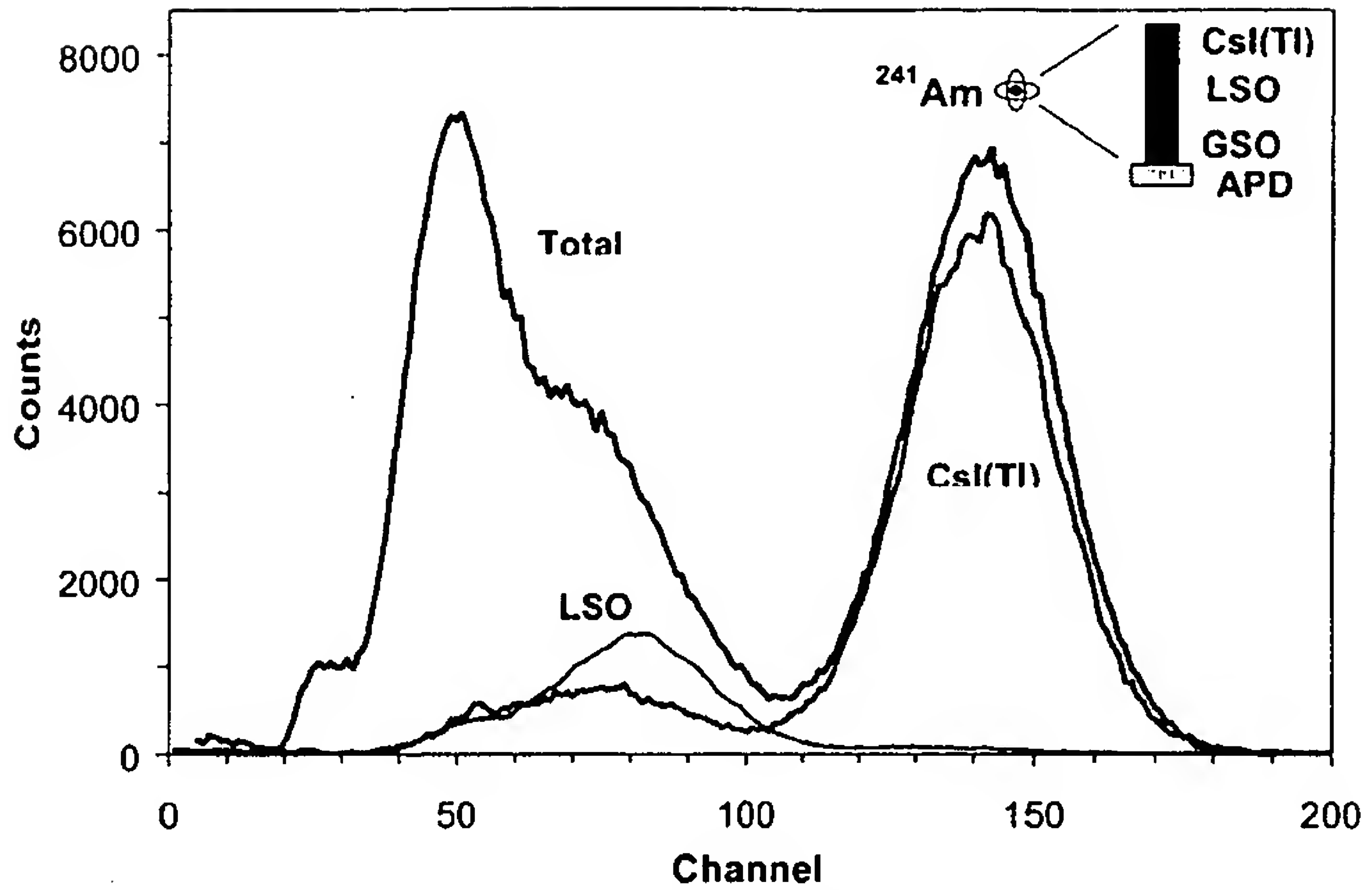


Fig. 7 – Total and PSD-gated energy spectra of ^{241}Am (60 keV) taken with detector assembly of Figure 1.

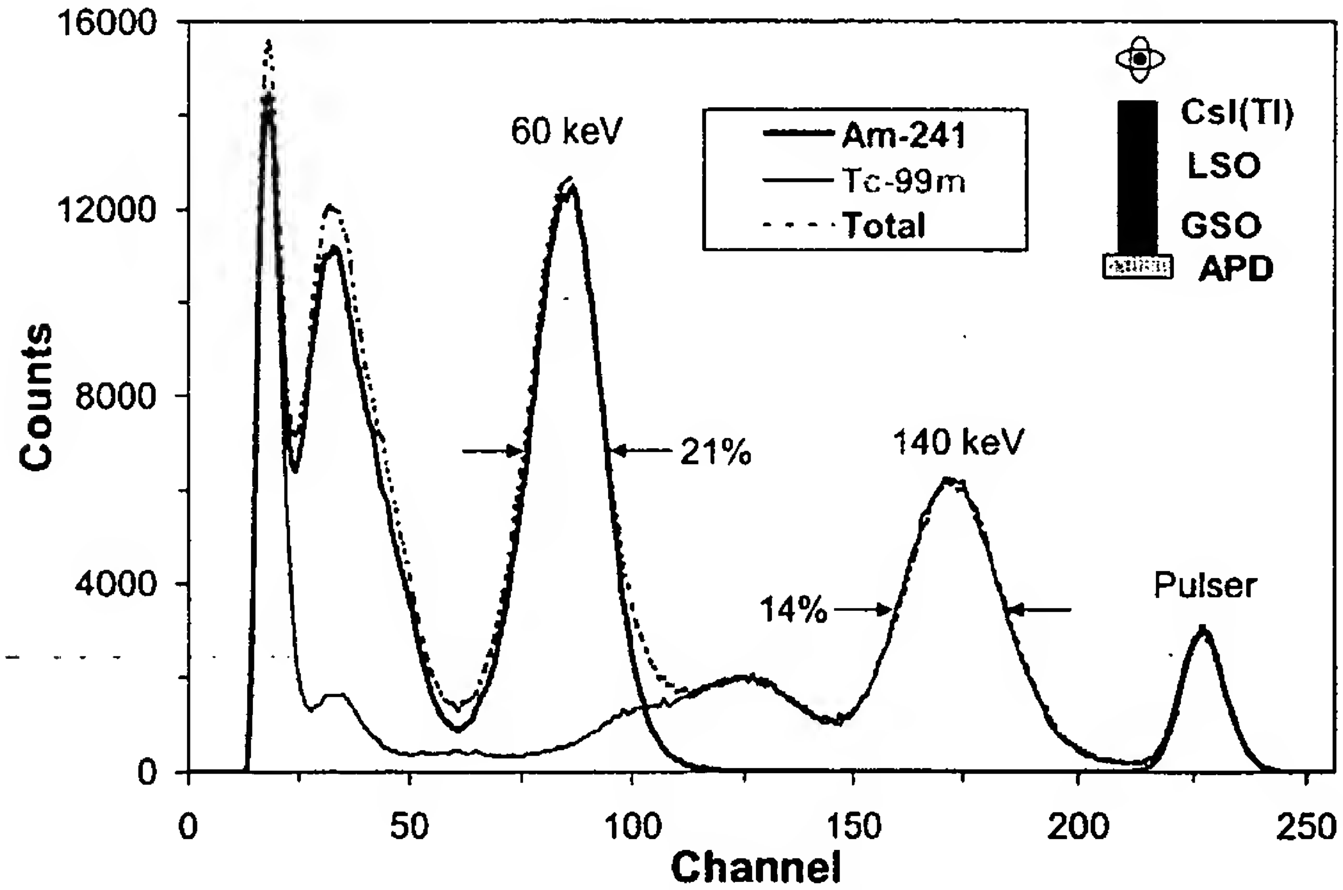


Fig. 8 – Energy spectra showing the separation between 60 keV (^{241}Am) and 140 keV (^{99m}Tc) that can be obtained in the front detector layer of CsI(Tl) with the detector assembly of Figure 1.